The Wavelength Selection for Calibrating Non-Contact Detection of Blood Oxygen Satuartion using Imaging Photoplethysmography

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***Abstract*—Based on imaging photoplethismography theory, two videos recorded at two different wavelengths of the visible light spectrum can be used to measure oxygen saturation (SpO_2), in a non-contact way. Here, we present better selection of these two wavelengths to detect SpO_2 level under ambient light and temperatures.**

I. INTRODUCTION

In recent years, detecting human physiological parameters including vital signs remotely and in a non-contact fashion has been a popular area of research. One of the most important human biological parameters is oxygen saturation (SpO_2) which shows the level of oxygen in blood and is a risk factor for chronic diseases of circulatory and respiratory system.

Traditionally, SpO_2 is measured in a non-invasive method using conventional pulse oximeters which are clipped generally to a subject's fingers or earlobes. This non-invasive method is suitable for many situations. However, there are circumstances where direct skin contact must be avoided, for example patients with serious burns or premature infants. Imaging photoplethysmography (iPPG) has solved this problem. According to photoplethysmography theory, which is based on blood absorption of visible light, the changes of blood in capillaries underneath the skin due to the cardiac pulse can be detected from very small variations in skin color. Visible light imaging devices such as cameras can be used to detect these small changes and extract the PPG signal [1].

II. DETECTION METHOD

A. Methodology and Challenges

Traditionally, the SpO_2 level, which indicates the intensity of oxygen in blood, can be expressed as

$$SpO_2 = \frac{HbO_2}{HbO_2 + Hb} \times 100\% \quad (1)$$

In order to detect oxygen saturation we need two PPG signals at two different wavelengths. In 2007, Kenneth Humphreys et al., for the first time, demonstrated a non-contact SpO_2 measurement system which detects two PPG signals at two wavelengths [2]. However, passive measurement (i.e. using ambient light) is more practical.

Where HbO_2 is oxygenated hemoglobin, and Hb is deoxygenated hemoglobin. Most pulse oximeters work based on the Beer–Lambert law and the theory of light transmission/reflection to extract SpO_2 from two PPG signals. According to the absorption spectrum of *HbO2* and *Hb* displayed in Fig. 1, if we choose two distinct wavelengths, we will be able to extract SpO_2 using Eq. 2

$$SpO_2 = A\frac{I_{AC}^{\lambda_1} / I_{DC}^{\lambda_1}}{I_{AC}^{\lambda_2} / I_{DC}^{\lambda_2}} + B = A \cdot R + B \quad (2)$$

Where I_{AC} represents the intensity of the AC signal from the light at two wavelengths caused by pulsation and I_{DC} denotes the maximum intensity of two beams of light. In Eq. 2, A and B are empirical constants determined by calibration. The two wavelengths should be selected such that the absorption coefficients of HbO_2 and Hb at one of the wavelengths differ greatly, but are approximately equal at the other wavelength.

Conventional finger-clip pulse oximeters usually use 660 nm and 940 nm wavelengths. Although these two wavelengths meet the requirements to extract SpO2, we cannot use these two wavelengths for iPPG based method which uses two cameras since the reflected light from the skin at these two wavelengths is very weak and we need to consider camera response. Therefore, we need to select two other wavelengths. In [3], Lingqin Kong and et al. used 520 nm and 660 nm filters for their experiments. One of the limitations of using a 660 nm filter is that the PPG signal at this wavelength is weak and noisy. The noise source of the 660 nm signal is not easy to predict. One reason is that the color of blood is red, which is close to 660 nm. Thus, the reflected signal from the blood is very strong, on the other hand, the variation of SpO_2 signal at 660 nm is very weak. The weak signal implies that even small noise sources will greatly reduce the signal to noise ratio of a measurement. Moreover, sun light has a very strong 660 nm component, overwhelming the PPG signal. These limitations make the 660 nm filter an impractical choice for all but the most controlled environments. Thus we decided to use a 460

nm filter instead of 660 nm filter. Therefore, we used two iPPG signals one at 520 nm and the other at 460 nm wavelengths to measure oxygen saturation.

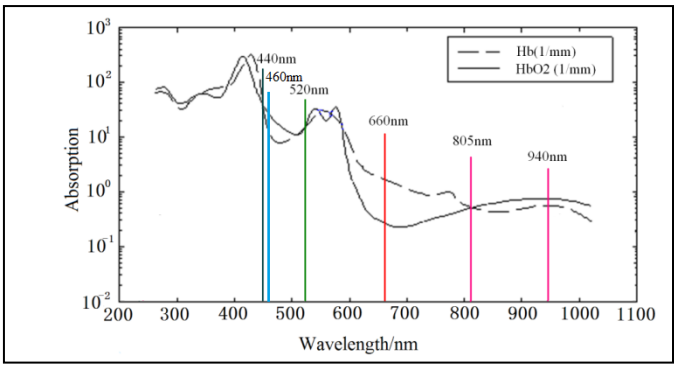

Fig. 1. The absorption spectrum of HbO_2 and Hb.

B. Experiments and Results

First, in the regular light of the room (standard lab fluorescent lighting), we setup two monochrome CCD cameras each mounted with a narrowband-pass (460 nm and 520 nm) filter about one meter from the volunteer. The cameras were focused on the volunteer's face. We select the area below eyes as the region of interest (ROI). We also use a finger-clip pulse oximeter (BVP sensor) as a reference. Data was captured from the oximeter and cameras simultaneously. The experiment was run for 20 seconds and video was captured at 40 frames/second.

Then we use an image processing algorithm implemented in MATLAB to extract PPG signals from each video. The algorithm we use first splits the ROI into lots of small regions, each region is 10 × 10 pixels. We average the value of each region and apply a 0.5Hz-5Hz bandpass to the signal. Finally, we sum up the effective signal and obtain the time (PPG) and frequency domain signals. From these PPG signals we are able to detect heart rate and respiratory rate. If we apply a bandpass filter to these PPG signals to remove the respiratory signal, we will obtain two clean PPG signals which are ready for oxygen saturation extraction as shown in Fig. 2.

In Fig. 2 we can see the time domain and frequency domain signals after de-noising and bandpass filtering. The red and black traces show the PPG signals at 520 nm and 460 nm, respectively. As we can see from the frequency domain signal, the two PPG signals have the same frequency. Next, we can extract I_{DC} and I_{AC} at each wavelength from the two PPG signals. The DC components are computed as the average value of the PPG signals over the corresponding periods of time. The peak-to-peak value extracted from the PPG signal in each cycle shows the AC component, which also varies when the blood saturation remains unchanged.

In order to accurately extract calibration coefficients (A and B) we need to conduct sets of experiments with a robust population of subjects at different ages and with varying blood oxygen levels measured from a reference sensor. After several groups of breath holding comparison tests with multiple volunteer subjects in the same environment temperature and light conditions, A and B were extracted.

$$SpO_2 = 128 - 44 \cdot R \quad (3)$$

The error in ambient light (about 500 lx) is 2% when the subject is at 1m distance from the two cameras.

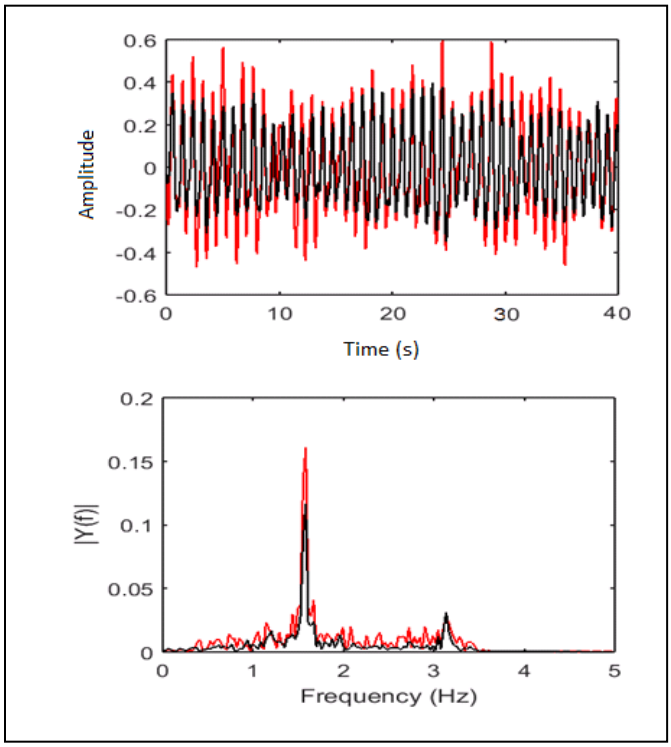

Fig. 2. The PPG signals after band pass filtering.

ACKNOWLEDGEMENT

This work is supported by the US Army Medical Research and Materiel Command under Contract No.W81XWH-14-C-0004.

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In the conduct of research where humans are the subjects, the investigator(s) adhered to the policies regarding the protection of human subjects as prescribed by Code of Federal Regulations (CFR) Title 45, Volume 1, Part 46; Title 32, Chapter 1, Part 219; and Title 21, Chapter 1, Part 50 (Protection of Human Subjects).

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